SUB-SAMPLED MOVING TABLE MRI

The invention relates to a magnetic resonance method for forming a fast dynamic image form a plurality of signals of an RF probe according to the preamble of claim 1. The invention further relates to a magnetic resonance imaging apparatus for obtaining a fast dynamic image according to the preamble of claim 11 and to a computer program

5   product according to the preamble of claim 12.

It is common practice to perform MR image acquisitions in which the patient position is physically moved through the magnet bore between a group of successive scans. At the end of this multi-station scanning, the resultant images are combined to a single large image. This approach is known e.g. from WO-A-02/04971, which allows to image a much

10   larger FOV than the limits of the magnet homogeneity, gradient linearity and RF coil uniformity normally allow. In this document a MR imaging method is described which involves the acquisition of sets of MR signals form several scan-volumes of an object. Different spatial approaches are taken in the scanning of the respective scan-volumes. In particular respective scan-volumes include different numbers of scan-slices or scan slices of

15   respective scan-volumes have a different slice thickness or scan-slices of respective scan-volumes have different FOV's (Fields Of View).

In US-A-5,636,636 a magnetic resonance method and device is described in which an object to be examined is displaced at a defined speed relative to an examination zone and a plurality of sequences act on the examination zone in the presence of a steady,

20   uniform magnetic filed. Each of the sequences includes at least one RF pulse and possibly a phase encoding gradient. An MR signal arising in the examination zone, after transposition to another frequency range by means of an oscillator signal, is used to produce an MR image. The method aims to enhance the quality of the MR image by preventing movement artefacts. This is achieved in that from one sequence to another, one or more of the frequency of the

25   RF pulses, the frequency of the oscillator signal and the phase position of the oscillator signal is adjusted in conformity with the position of the object to be examined relative to the examination zone so that a part of the object which is imaged in the MR image moves relative to the examination zone in synchronism with the object.

In EP-A-1 024 371 a magnetic resonance imaging apparatus is described in which excitation pulses are applied to a restricted region of the bore of the MR imaging magnet in which the magnetic field is uniform. The data samples collected are Fourier transformed to form a volumetric image of the restricted region. A motor continuously moves

5   a patient couch so that a region of interest passes through the region of the good field. The collected data samples are corrected to compensate for the motion so that a volumetric image is formed of greater length than that of the restricted region.

Since all the above-mentioned methods require a full scan within the restricted FOV imaging of the full object will take quite a long sampling time

10   It is an object of the present invention to provide a method for forming an MR image from data sampled in an array of adjacent FOV's in shorter time. It is a further object of the invention to provide an adequate apparatus and a computer program product for exercising the method.

This and other objects of the invention are accomplished by a method as

15   defined in claim 1, by an apparatus as defined in claim 6 and by a computer program product as defined in claim 7.

It is a main advantage of the present invention that images with suitable large FOV's are obtained with a short magnet in shorter time.

These and further advantages of the invention are disclosed in the dependent

20   claims and in the following description in which an exemplified embodiment of the invention is described with respect to the accompanying drawings. It shows

Fig. 1 a schematic view of a patient at three different positions moved through

25   the bore of a main magnet for MR imaging,

Fig. 2 the data for each table position of Fig. 1,

Fig. 3 the virtual sensitivity map for the full FOV,

Fig. 4 a schematic representation of the final image and FOV after reconstruction,

30   Fig. 5 an arrangement with a separate array of body coils mounted on the patient table,

Fig. 6 prior art imaging showing data from two separated stations of the phantom and the combination of both data sets,

Fig. 7 prior art imaging showing data measured by the same stations as in Fig. 6 with a band limiting filter in the measurement (frequency encoding) direction,

Fig. 8 a first embodiment according to the present invention showing a phantom scanned at 3 different positions moved through the bore of the main magnet of an MR imaging system,

Fig. 9 a second embodiment according to the present invention, wherein data is measured as in Fig. 6 with a SENSE factor of 1.33 at each station,

Fig. 10 the different steps for decoding the data according to the present invention, and

Fig. 11 a third embodiment according to the present invention, wherein data is measured at three separate stations, with different SENSE factors and its reconstruction.

Specific numbers dedicated to elements defined with respect to a particular figure will be used consistently in all figures if not mentioned otherwise.

The expression "antenna" is used as a more general term for transmitting and receiving coils. The sensitivity encoding method called "SENSE" as used in the present invention has been developed by the Institute of Biomedical Engineering and Medical Information, University and ETH Zürich, Switzerland. The SENSE method is based on an algorithm which acts directly on the image as detected by the coils of the magnetic resonance apparatus and which subsequent encoding steps can be skipped and hence an acceleration of the signal acquisition for imaging by a factor of from two to three can be obtained. Crucial for the SENSE method is the knowledge of the sensitivity of the coils which are arranged in so called sensitivity maps. In order to accelerate this method there are proposals to use raw sensitivity maps which can be obtained through division by either the "sum-of-squares" of the single coil references or by an optional body coil reference (see e.g. K. Pruessmann et. al. in Proc. ISMRM, 1998, abstracts pp. 579, 799, 803 and 2087).

In the SENSE technique it is usually required that at least two RF receiving coils are present. Different sensitivity maps between the two RF coils for the same imaging position is a prerequisite for the SENSE method. This requirement is usually achieved by placing the RF coils at physically different locations with respect to the region of interest. With the SENSE method and two different RF receiving coils the number of phase encoding

steps can be reduced, and consequently also the amount of acquired data, by a factor of two.
This is particularly advantageous with respect to reduction of imaging time.

If only a single RF receiving coil is present the usual approach to SENSE
would not work. However, if a single coil is positioned at two different locations two inde-

5   pendent partially encoded acquisitions at the same position of the imaged object can be made,
so that the SENSE method can be applied. The data available at the end of such a scan is
identical to the usual SENSE acquisition except that acquisition time is twice as long, which
is as long as normal acquisition without SENSE. The advantage of this acquisition method is
that a larger FOV is obtained than either the RF receiving coil or the magnet homogeneity,

10 gradient linearity and RF transmitting uniformity allow. This situation exists in the case of
magnets with a short imaging volume along the Z axis.

In figure 1 the contours of a main magnet 1 with a magnet bore 2 is
schematically depicted. A patient 3 on a movable table 4 can be moved through the bore 2 in
discrete steps, here at three different table positions in which the abdomen of the patient is

15 scanned (Fig. 1a), the breast of the patient is scanned (Fig. 1b) and the head of the patient is
scanned (Fig. 1c). Within the main magnet 1 there are mounted a transmitting quadrature
body coil 6 and a smaller receiving quadrature body coil 7. The receiving coil 7 is defining
the dimensions of the Field-of-View (FOV) of the image. In this example data with the
restricted FOV 8 of the receiving coil 7 are sampled, in order to form a single MR-image of

20 the entire region of interest or full FOV 9 as indicated by the dashed lines. Arrow 10
indicates the encoding and/or foldover direction. The subsequent data sampled at the three
table positions are reconstructed by the SENSE method. In figure 2 the images at each table
position encoded for the full FOV 9 is shown, whereas different fold-over artefacts are
obtained from each different scan. In figure 3 the virtual coil sensitivity map for each of the

25 table positions are shown, which is actually a triplicate of the single sensitivity map of the
single receiving coil 7. From the sensitivity map of the full FOV (Fig. 3) an unfolded image
can be reconstructed as shown in a schematic representation of the final image and the full
FOV. The total number of encodings acquired is just the same as if a fully encoded scan
would be possible on the full FOV, here 3 * N pixels as N encodings are provided for the

30 restricted FOV.

In an alternative embodiment it is possible to utilize a single coil for
transmission of RF signals and for receiving RF signals. In this case the sensitivity
information required for the SENSE reconstruction must be supplied by an alternative means
such as calculation of the theoretical sensitivity behavior.

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Another embodiment of the present invention is shown in figure 5 in which an array of local surface coils 11 are mounted at a fixed position relative to the patient 3, i.e. relative to the table 4. Thus, these coils 11 are moved by the table movement and only the ones which are positioned within the restricted FOV of the transmitting coil 6 are activated to

5   receive the transmitted RF signals. In this case the sensitivity maps of the coils 11 are needed to reconstruct the final image by the SENSE method.

The above mentioned approach has particular advantages in the case of short main magnets with a restricted nominal FOV, which enables a much larger FOV to be imaged in a shorter time than would otherwise be necessary. For this reason shorter magnets

10   can be used which is especially advantageous for patients with a tendency to claustrophobia.

Although the method is described with three different table positions it will be clear to the skilled person that the method can already be applied with two different positions. Otherwise, a much larger amount of table positions can be applied too. With suitable receiver coils it is also possible to combine this approach with the conventional SENSE method singly

15   or simultaneously along both the direction of the table movement and any other orthogonal direction.

Figure 6 illustrates a prior art example of imaging with two stations. When the measurement (phase encoding) gradient is orientated along the direction of the table movement in feet to head direction (FH), it is necessary to acquire a larger FOV at each

20   station so as to avoid image folding. The extra data is then discarded at the places indicated so that the image sections from the two stations can be combined. In this example, each station requires 450 encodings in order to generate a full FOV equivalent to 675 encodings. So, a total of 900 encodings are required to generate a full FOV equivalent to 675 encodings. This represents an additional scan-time overhead of 25%.

25   Figure 7 illustrates an alternative prior art example of imaging with two stations with a band limiting filter in the measurement (frequency encoding) direction. In this example, the measurement (frequency encoding) gradient is orientated along the direction of table movement (FH). The advantage of doing this is that the data, at the point of image combination, can either be bandwidth limited during acquisition or simply thrown away after

30   reconstruction. Since this is the frequency encoding direction there is no time penalty. However, in order to avoid folding in the left to right direction (LR), it is still required to oversample along the encoding direction to the full FOV. Once again, in this example, each station requires 450 encodings, and 900 in total, to realize a full FOV equivalent to 675 encodings. The additional scan-time overhead of 25% remains.

Figure 8 represents a first embodiment of the invention showing a phantom scanned in three different positions moved through the bore of the main magnet of an MR imaging system. In order to decode the information from the data obtained by the SENSE scan at different adjacent positions as shown and described with respect to Fig. 1 additional

5   information is needed from the neighboring scans. This can e.g. be obtained by providing overlapped scans as can be seen in Fig. 8a. Suppose that the full FQV ranges over slices A to I and the scans are made with a SENSE factor of 1.66. Then for the left scan the information of slices A and E are folded-in in slices D and E respectively, for the middle scan the information of slices C and G are folded-in in slices D and F respectively, and for the right

10   scan the information of slices E and I are folded-in in slices H and F respectively. The adjacent regions D and F can be unfolded since the sensitivity maps of RF coil 7 from the position of the left image and the middle image, and from the middle image and the right image respectively are known. After unfolding half of the region D and F will be discarded (see Fig. 8b), or corrected and co-added, in order to obtain a contiguous image in these

15   regions. From the unfolding in region D the unfolded information of region A can be obtained, so that the full information of region A by the SENSE method will be recovered. In the same manner region I can be unfolded and so the complete image over the full FOV will be obtained (see Fig. 8c). In this embodiment, N encodings are acquired at each of the 3 stations in order to realize a full FOV of 3*N pixels.

20   Figure 9 shows the second embodiment which illustrates how the invention provides an advantage over the prior art described in Fig 6 and 7. In this example, the encoding direction is orientated along the direction of table movement. Instead of acquiring 450 encodings per station, 340 encodings are acquired and the image is allowed to fold. The position where folding appears can be controlled during the acquisition by applying a phase

25   increment to the receiver demodulation frequency for each encoding step. In this manner, the folded region is moved from the center edge of the image from each station to the outer edge. The same effect can be achieved after reconstruction by scrolling the images. The lower image of figure 9 illustrates the combination of the images from both stations. The regions B, C, D and E can be easily connected. However, the regions A and F are folded with regions D

30   and C respectively. In the acquired data representation (top diagram), the regions A and D are folded together. In actuality, these regions are acquired at the periphery of the homogeneous region of the imaging volume and so it is likely that regions A and D will be geometrically distorted due to the poor homogeneity in those regions. This fact is indicated by the apostrophe in A' and D'. Where A' represents a geometrically distorted version of region A

and D' a geometrically distorted version of region D. The same is true for the second station data where region C is folded in region F. Here, the presence of geometric distortion is indicated by the apostrophe in F' and C'. To unfold the data in these two regions it is necessary to take the geometric distortion into account.

Figure 10 illustrates the steps required to unfold A' and D' while taking the geometric distortion into account. It should be understood that, when the magnet homogeneity and gradient coil linearity are known, image geometric distortion can be corrected by warping or morphing the pixels back to a calculated correct position. The converse is also true i.e. a given portion of an image can be pre-distorted in a manner similar to the effect that a known magnet homogeneity and gradient coil linearity would produce. This reciprocal relationship is used here. With reference to figure 10, the image section A'+D' represents the acquired data including geometric distortion, from station 1 of figure 9. The image section D represents the acquired, undistorted, data from station 2. With knowledge of magnet homogeneity and gradient coil linearity it is possible to pre-warp D and, after suitable masking, generate D*. D* is essentially equivalent to D' and can be used in combination with the acquired A'+D' image section to separate out the region A'. If desired, the separate A' region can be further corrected (shown in figure 9) to remove the effects of geometric distortion before combining into the full image. With reference to figure 9, the same approach can be used to separate F' from F'+C' to generate F. This is all possible because each of the image sections involved are acquired while the table top is at different positions with respect to the sensitivity profile of the receiving coil. Image regions A'+D', D, C and F'+C', by virtue of the multi-station acquisition, are each acquired at a uniquely different region of the receiving coil sensitivity.

Figure 11 describes a further embodiment in the case of a 3 station acquisition. In this embodiment, the different stations have different SENSE factors. The two stations either end of the full FOV use a SENSE factor of 1.33 and the middle station uses a SENSE factor of 2.0. Compared to the alternative acquisition, a total of 910 encodings are acquired versus the alternative of 1350. This represents a total scan-time saving of 33%. Regions F'+D and F can be used to determine D. Region D, together with region A'+D', can be used to determine region A. Likewise, region E+C' and C can be used to determine region E. Region E, together with region H'+E can be used to determine region H.